INJURY BIOMECHANICS RESEARCH
Proceedings of the Twenty-fifth International Workshop

Biomechanics of Cervical Spine Hyperflexion Injuries

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ABSTRACT

Hyperflexion injuries of the human cervical spine commonly occur in vehicular crashes, contact sports, diving, and falls. Flexion related injuries constitute a high percentage of all cervical spine injuries. The objective of this study was to determine the mechanisms and tolerance of the human cervical spine under hyperflexion loading conditions through *in vitro* biomechanical experimentation. The impact load was delivered by an electrohydraulic piston to the cranium of a cadaver head-neck complex in a preflexed configuration. Measured biomechanical quantities included head impact force, neck forces and moments, kinematics, spinal cord pressure, and pre-test alignment parameters. Of the eight specimens tested, five had disruption of the posterior ligaments in the C5-T1 region while the other three had severe hyperflexion disruption of one entire lower joint in the C4-C7 region with or without associated vertebral fractures. The anterior eccentricity of the occipital condyle with respect to T1 body was the most critical variable which influenced the loading condition and injury outcome. The spinal cord pressures were consistent with the severity of joint disruption and moment magnitude. The human tolerance for hyperflexion injuries is suggested as 60 Nm for minor trauma and 85 Nm for severe trauma.

INTRODUCTION

Hyperflexion injuries of the human cervical spine commonly occur in vehicular crashes, contact sports, diving, and falls. Hyperflexion of the cervical spine is a forced forward bending exceeding the normal physiological range. The resulting trauma may include those classified in the literature as hyperflexion, flexion, compressive flexion, and distractive flexion injuries [1, 5, 14]. Epidemiological studies have indicated that the flexion mechanism is the most common cause of neck trauma and accounts for 48-70% of all cervical spine injuries [1, 14].

Hyperflexion injuries are often associated with localized ligamentous disruption of the cervical spine [5]. They may be characterized by localized kyphotic angulation at the level of injury, anterior dislocation or rotation of the subluxed vertebra, anterior narrowing and posterior widening of the disc space, dislocation of the articulating facets, or widening of the space between spinous processes ("fanning") [4, 5]. In the absence of dislocation, flexion injuries are difficult to detect by radiographic means since there are often no bony fractures. The injured cervical spine may appear normal in radiographic films when the spine is in neutral or extended positions. The radiographic signs of injury are only evident in flexion. Such injuries may lead to late dislocation or late instability of the vertebral column with potentially disastrous consequences [4]. These injuries are difficult to stabilize with non-operative treatment [12].

Our previous experimental studies have demonstrated that the pre-alignment condition of the head-neck complex significantly influences the injury mechanism and biomechanical response variables [7, 9, 11]. Determining the correlation between the alignment condition, biomechanical responses, and injury mechanism may assist in the evaluation of injury risks and preventive measures. The present *in vitro* experimental study was designed to biomechanically determine the injury mechanism and tolerance for hyperflexion injury of the cervical spine under head impact.

METHODS

The dynamic experiments were performed on human cadaver head-neck complexes using similar techniques according to a previously established protocol [7]. Specimens were tested with ages ranging from 39 to 61 years. The fresh head-neck complexes were isolated from the human cadaver at the T2-T3 disc. Medical records and radiographs were examined to exclude specimens with pre-existing pathological conditions or anatomic defects. The posterior muscles and skin were

kept intact while those on the lateral and anterior sides were dissected. The inferior end of the specimen, along with the skin, was fixed with polymethyl-methacrylate (PMMA) while leaving the C7-T1 joint and above unrestrained. Retroreflective pin targets were placed in the anterior vertebral body, the lateral vertebral masses, and the cranium. The spinal cord pressures during loading were recorded by seven pressure sensors attached to an instrumented artificial spinal cord developed in another study [8].

The inferior mount of the preparation was then bolted to a six-axis load cell and firmly attached to an x-y cross table of the custom-designed servo-controlled electrohydraulic piston. The specimen was flexed prior to loading so that the head impact would produce a hyperflexion of the cervical spine (Figure 1). A lateral radiograph was taken prior to the impact to record the geometric relations of the head, vertebrae, targets, and the impactor. A flat plate with 6 mm thick ensolite padding attached to the piston of the testing device served as the impact surface. A vertical impact load was delivered to the head at 2 to 5 m/sec by the electrohydraulic piston.

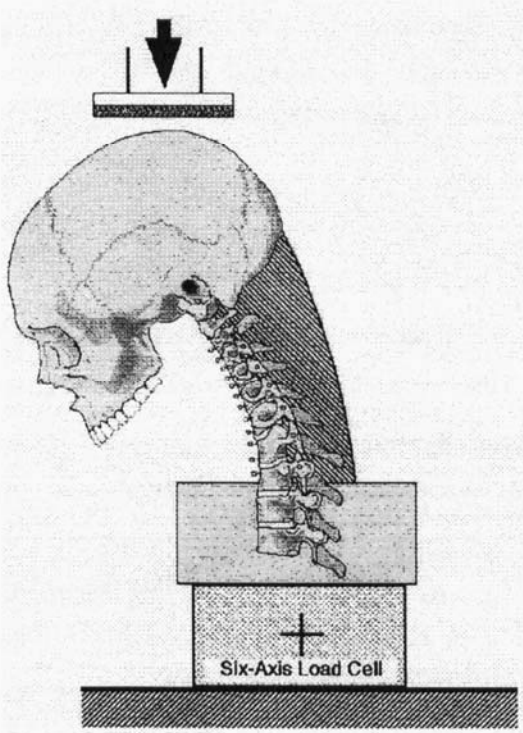

Figure 1. Schematic diagram of experimental set-up for *in vitro* hyperflexion testing.

Transducer measurements included an in-series load cell and displacement gauge in the piston. The six-axis load cell recorded the generalized force histories at the inferior end of the preparation. The transient spinal cord pressures were recorded by the sensors at the seven cervical vertebral levels. The kinematic responses of the head-neck in the sagittal plane were recorded from the lateral perspective at a rate of 4500 frames per second by a high-speed digital video camera system. The radiographs of the specimen in flexion and extension were taken for examination of the injuries after the test. The specimen was visually examined for tissue damage immediately after the test. Visible and radiographic injuries to the cervical spine were identified and classified by a spine surgeon.

The biomechanical analyses for the present study included the eccentricity, head impact force, neck resultant force, sagittal bending moment at the injury level, and spinal cord pressures. The eccentricity was defined as the horizontal distance between the center of the occipital condyles and the T1 posterior body margin and measured from the pre-test lateral radiograph. The neck resultant force was obtained from the six-axis load cell data by vector summation. The sagittal bending moment at the site of injury was obtained using the load cell forces and moments and their geometric relations based on principles of mechanics. The geometric relations of the forces and moments were measured from pre-test radiographs and high-speed video images. Using a free-body diagram and the equations of equilibrium, the bending moment at the injured level was calculated.

RESULTS

The hyperflexion injuries produced in this study were classified into three groups: (I) disruption of the C7-T1 posterior ligament complex; (II) flexion ligamentous injury with vertebral fractures at higher levels than the injured ligamentous structure; and (III) primarily severe disruption of one entire joint with or without associated vertebral fracture. The three injury groups exhibited different cervical spine deformation patterns in the sagittal plane as revealed by the analysis of the high-speed video. In the first group, the deformation of the spine was a continuously forward rotation of the cervical vertebrae with respect to T1 resulting in extreme hyperflexion of the C7-T1 joint. Specimens in the third group experienced initial compression of the entire column followed by local hyperflexion at the injured level other than the C7-T1 joint. The deformation behavior of the second group had characteristics of both.

The resulting injuries depended on the initial specimen alignment condition which was

measured in terms of eccentricity. The eccentricities of all the tests ranged from 3.5 to 11.2 cm. The minor injury group (I) had greater eccentricities (9 to 11 cm). The severe injuries (group III) were associated with much smaller eccentricities (3 to 4 cm). The injury group II specimens had intermediate eccentricities.

The injury outcome was directly related to the load magnitude and load component (axial force and bending moment) in the cervical spine. The peak axial head impact force ranged from 3000 to 9700 N. The peak neck resultant force ranged from 750 to 5250 N. The peak flexion bending moment at the injured level ranged from 50 to 90 Nm. Group I specimens sustained low magnitudes of force and moment in the cervical spine with an average of 1115 N (750 to 1450 N) and 61 Nm (52 to 70 Nm), respectively. Group II specimens sustained increased forces with flexion injury moments at similar magnitudes (50 to 70 Nm)a. Group III sustained high moments (75 to 90 Nm) at the injured level with moderate to high cervical spine forces (2368 to 3623 N).

The spinal cord pressures were consistent with the magnitude of bending moment. The pressures at uninjured levels were of low or zero magnitude while it was generally higher at the level of injury. The cord pressures at or near the level of injury were generally below 20 N/cm^2 for groups I and II specimens. Much higher magnitudes of cord pressures (45 to 100 N/cm^2) were recorded at or near injured levels for group III specimens.

DISCUSSION AND CONCLUSION

The present study is an extension of our earlier research which focused on the biodynamics of compression injuries of the human cervical spine [7-11, 15, 16]. The present study has focused on flexion-related injuries of the cervical spine. Although the methodology used is similar to that of the previous studies, the loading vector in the cervical spine in this study was primarily flexion bending moment rather than the axial compressive force. The resulting injuries were flexion-related joint disruption with or without vertebral fractures. Injury outcomes were correlated to pre-test alignment conditions, vertebral kinematics, neck forces and moments, and spinal cord pressures.

The injuries produced in this study are consistent with those described in the literature on hyperflexion, flexion, compressive flexion, or distractive flexion injuries of the cervical spine [1-6, 13, 14]. Clinical data suggest that hyperflexion injuries occur primarily at single levels [5, 14]. Adjacent injuries tend to be relatively minor posterior element fractures or ligament avulsions. This is most

likely because extreme flexion of the cervical spine is highly localized at a single level although the entire cervical spine may bend in flexion in general, as confirmed by the analysis of the high-speed video images.

The results of this study demonstrate that the pre-test alignment condition, in terms of the eccentricity, is a critical biomechanical variable for injury outcome. A large eccentricity tends to result in low force and simple ligamentous hyperflexion injuries. The group I specimens had larger eccentricities and sustained disruption of posterior ligament complex with low neck loads. With decreasing eccentricity, the force sustained by the cervical spine increased, as did the risk for vertebral body fractures. Severe flexion injuries with high bending moment were produced only in group III specimens with small eccentricities. This was consistent with local extreme flexion at the injured levels as confirmed from the high-speed video images.

The spinal cord pressure responses are consistent with moment magnitude and severity of osteoligamentous injury. The spinal cord pressures for groups I and II specimens with relatively low moments were mostly well below 20 N/cm^2 through the entire cervical spine. The spinal cord at such low levels has minimum risk for acute injury [8]. This agrees with the literature that neurologic damage in minor hyperflexion injury is usually mild and reversible [1, 3]. The cord pressures for group III specimens with high bending moments had much higher spinal cord pressures at or near the injured level indicating increased risk of cord injury. In particular, one specimen had a tear-drop fracture with tremendous spinal deformations and recorded the highest cord pressure (100 N/cm^2). This result agrees with the clinical observation that flexion tear-drop fractures may be characterized by acute anterior cervical cord syndrome with immediate complete quadriplegia [1, 6].

The sagittal bending moment at the injured level was found to be the most consistent biomechanical variable which quantified the severity of hyperflexion injuries. Flexion moment in the cervical spine induces tension in the posterior ligament complex and compression in the anterior portion of the vertebral bodies and discs. Injury of ligaments are caused by tensile loading while vertebral body fractures are usually caused by compressive loading. Thus, a high flexion moment with low compressive force causes tensile failure of the posterior ligamentous tissues and progressive bending with increasing moment magnitude. Simple ligamentous injury can be measured by the bending moment alone. Groups I and II with relatively minor injuries in the present study fall into this category. Based on the average moment for groups I and II, a suggested tolerance for minor

hyperflexion injuries with minimum risk for cord damage can be set at 60 Nm. A suggested tolerance for the more severe trauma based upon the remaining specimens is 85 Nm. This study offers initial parameters useful for defining injury tolerance of the human cervical spine to hyperflexion.

ACKNOWLEDGMENT: This study was supported in part by PHS CDC Grant RC49CCR-507370, and the Department of Veterans Affairs Medical Research.

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DISCUSSION

PAPER: **Flexion Injuries of the Cervical Spine**

PRESENTER: Frank Pintar, Medical College of Wisconsin

QUESTION: Barry Myers, Duke University

Frank, nice study! I'm a little concerned about what these initial and boundary conditions represent. In particular, the one catastrophic injury that you showed has a retrolisthesis, where the top vertebra is behind the lower one. That is not something you see in flexion injuries. The hallmarks are antrolisthesis.

A: Right.

Q: These subjects are flexed, tipped over, protracted and then clonked on the back of the head. I just wonder what that represents.

A: We are trying to produce the kinds of injuries that are seen in clinical environments and granted, the resting state of that one isn't what you would see in an x-ray. However, cadaver x-rays can never be like what you see in terms of the in vivo resting state because you have the muscle response that occurs afterwards. In fact, I think some of these flexion injuries that we see in the patient population have a lot to do with muscular response after the injury occurs so the retrolisthesis is certainly not representative of the clinical population because there is some muscle response that goes on afterwards.

Q: That is certainly true. How a patient presents doesn't necessarily look like how they did at the time of injury. On the other hand, when we produced facet dislocations in compression/flexion, either statically or dynamically, they really do anterolisthese and so, I'm still a little nervous about it.

A: Clinically we see subluxations occurring usually at just one level and with your conditions, from what I remember, you produced other injuries superior to that. So, I am equally concerned with the way that we are producing these injuries. Maybe we can't really do it in this environment.

Q: Maybe we'll just disagree, but dynamically when you get a facet dislocation, the top vertebrae goes over the bottom.

A: Agreed, but in clinical cases you only see that injury. You don't see other injuries.

Q: But also in the lab you see that. So, when you show a retrolisthesed vertebra, I'm concerned. That is a strange way to hit the back of your head and I don't know how often people do that in the real world.

A: Like I said, the patient will never end up like that because his muscles will want to stabilize his spine.

Q: We should just disagree. They might not end up that way because they don't hit their heads that way, so it might be worth trying some different ways to hit them.

A: Our clinicians almost always say that they see bruises on the back of the head with these injuries.

Q: Guy Nusholtz, Chrysler

My understanding from what you said was that you were estimating the moments just by figuring out what the eccentricity was and then doing the calculation from the load cell. Did you look to see what effect the differential motion in the different vertebrae were going to have on that moment estimation? Did you do some sort of error calculation?

A: Yes, we looked at that in terms of how much the failure site can move before it will make a difference. It is valid until you get to about the peak, but after the peak there could be a lot more bending that goes on. Since that could affect the calculation, the moment trace may not be valid after the peak.

Q: OK.

Q: Roger Nightingale, Duke University

Frank, one thing, when we've been calculating our moments in our impact tests, we found that the calculations are very sensitive to where exactly in the motion segment you choose to calculate it. When you have a 2000 Newton compressive load, plus or minus one centimeters, that's 20 Newton-meters, one way or the other. Did you choose a specific site, like anterior body or posterior body?

A: Generally we used mid-body at the failure site, but you're right, it does depend on what location you choose.

Q: It is a tough problem. Maybe this is something that we should all think about and try to standardize, because we really don't know where it should be calculated necessarily.

A: Agreed.